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Aging of Bone Tissue: Mechanical Properties*

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ABSTRACT: The mechanical properties of machined cortical bone specimens from human femora and tibiae were determined in tension, torsion, and compression for a population ranging in age from twenty-one to eighty-six years. No significant differences were found between the mechanical properties of male and female specimens. Tibial specimens had greater ultimate strength, stiffness, and ultimate strain than femoral specimens. Consistent decreases with age for all mechanical properties except plastic modulus were found in the femoral but not in the tibial specimens. No consistent significant differences in tension properties were found in specimens from normal, osteoporotic, and corticosteroid-treated individuals.

Although the question of the change in mechanical behavior of bone tissue with increasing age has been the subject of several studies, the results have been equivocal. Melick and Miller reported that the tensile strength of adult human femoral cortex declines approximately 4 per cent per decade, using a slow strain rate that produced fracture in two minutes. Since the mechanical properties of bone tissue are dependent on the rate at which loads are applied, these results are difficult to interpret in relation to the usual rates of loading that occur during traumatic injuries. A similar rate of decay of the shear strength was found by Hazama, his values for the sixty to eighty-year-old group being 85 per cent of those for the twenty to twenty-nine-year-old group. In contrast, Evans and Lebow, using embalmed femoral samples, found no decrease in tensile strength with age. Sedlin and Hirsch, using a bending test on fresh human femoral samples, also found no trends for tensile strength with increasing age. The work of Vose and associates with embalmed senile osteoporotic femora showed that specimens of this tissue had a higher bending strength than specimens from the normal population, but that the bending strength of the osteoporotic whole bone was decreased as expected. All previous studies of age-related changes, however, were conducted before the importance of the elastic-plastic characteristics of bone tissue was demonstrated¹. In the study reported here we examined the elastic-plastic failure characteristics of cortical bone tissue from both the tibia and the femur of subjects in the third through the ninth decade of life.

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Materials

Two sources of cadaver material were utilized. Through a cooperative program, the Department of Orthopaedics, Catholic University of Louvain, Belgium, provided thirty-three pairs of specimens of fresh frozen human femora and tibiae ranging in age from twenty-nine to eighty years. At the time of death, a physician took histories from both the family and the private physician. Since Louvain has a non-mobile population, all the subjects had been known to their private physicians for extended periods of time. Information regarding their physical activity usually engaged in, activity prior to death, and illnesses and diseases was obtained in all cases. One femur and one tibia for each subject were used for material testing. The other bones were used in another study still in progress to determine whole bone strength.

The second source of cadaver material was the United States Navy Bone Bank. Bones which could not be used for implantation because of contamination were frozen immediately after autopsy. These specimens ranged in age from twenty-one to eighty-six years.

Combining these two groups of specimens, we had four from the third decade, two from the fourth, four from the fifth, eight from the sixth, ten from the seventh, nine from the eighth, and two from the ninth decade. Most of the individuals whose bones were used had engaged in either heavy labor or normal to vigorous activity. The periods of decreased activity prior to death ranged from one month to seventeen years. Seventeen had been bedridden for from one month to four years prior to death.

Methods

The bone segments were wrapped in cloth, saturated with physiological saline, covered with impermeable plastic, and frozen at temperatures not higher than -20 degrees centigrade. The material was shipped in insulated containers packed with dry ice. Techniques of specimen preparation were those we described previously⁹. Briefly, high-speed milling of specimens under constant water drip was used to prepare specimens of two shapes: square-cross-section bars and square-cross-section dumbbells³. The dumbbell-shaped specimens, used for tension and compression tests, were cut from the diaphysis with their long axis oriented parallel to the long axis of the bone. The center portion of all specimens coincided with the middle of the thickness of the cortical wall. Thus, no specimen contained bone tissue which was immediately adjacent to either the endosteal or the periosteal surface. The square-

cross-section bars were used for the torsion tests. All tests were carried out at deformation rates which produced fracture approximately 0.5 second after the onset of loading.

The testing machine for the modes of tension and compression was described previously⁹. Loads were measured with a strain gauge load cell which was used for both tension and compression tests. Strains for both tension and compression tests were determined with a clip-on extensometer. This technique of strain measurement was chosen since it did not require surface drying or alteration which would be necessary if strain gauges were used. Testing instrumentation allowed the surface of the specimens to be wet throughout the entire test so that plastic deformation would not be suppressed².

Torsion tests were performed in a standard torsion test apparatus¹ which simultaneously determined the transmitted torque and the imposed angular deformation. Specimen handling techniques were similar to those used in the tension and compression tests.

All load-deformation data were recorded by photographing a storage oscilloscope on which load (or torque) and extensometer reading (or angular deformation) had been recorded.

It was assumed that bone tissue displays tensile characteristics which can most accurately be modeled by an elastic-plastic description. Therefore, yield stress (above this stress the deformation no longer returns to zero when the load is released), ultimate stress (stress at the point of fracture), ultimate strain (strain at the point of fracture), elastic modulus (slope of the reversible or elastic portion of the stress-strain curve), and plastic modulus (slope of the irreversible or plastic portion of the stress-strain curve) were defined for tension tests as illustrated in Figure 1. The points of the load-deformation curve were transformed to points on a stress-strain curve by dividing the load coordinates by the initial area of cross section of

the specimen and the deformation coordinates by the initial length. No attempt was made to make allowance for any of the void spaces normally found in bone. The energy absorbed to fracture for each specimen was calculated as the area underneath the tension load-deformation curve. Since all specimen dimensions were nearly identical with no more than a ± 2 per cent difference, total energy for each specimen was not normalized with respect to the volume of the specimen.

The compression elastic modulus was calculated from the initial slope of the compression load-deformation curve while the ultimate stress was calculated by dividing the highest load sustained by the initial area of cross section. The shear modulus (the ratio of the induced shear stress to the resulting shear strain) was calculated from the initial slope of the torque-angular deformation curve.

Results

Tension Properties

Tension tests were performed on thirty-three femora and twenty-eight tibiae, twenty-four of the femora and twenty-four of the tibiae being paired bones from twenty-four individuals. The tensile mechanical properties by decade are shown for femoral and tibial bone tissue in Table I. To determine whether the properties of tibial bone differed from those of femoral bone, the mean values for each of the five tensile properties (yield stress, ultimate stress, elastic modulus, plastic modulus, and ultimate strain) for each of the twenty-four bone pairs were compared by calculating the ratio of each tibial tissue property to the corresponding femoral tissue property. The results for the femora and tibiae are compared in Table II. The mechanical properties of tibial tissue were significantly greater except for ultimate strain. Because of this, the tibiae and femora were treated as two separate groups for the rest of the analysis of tensile properties. No significant difference ($p > 0.05$) was found for any material property when age-grouped specimens from males and females were compared using the t test.

The tensile mechanical properties of femoral tissue generally decreased with age, with the exception of the plastic modulus. When the mean values for the bone properties of the femur in each decade were subjected to linear regression analysis, high correlation coefficients ($r > 0.8$) were obtained for all properties except the elastic modulus. In addition, the values of the properties for the third decade were compared with those for the ninth decade and all except those for elastic modulus and energy absorption were different ($p < 0.01$). The larger per-decade changes were in the plastic properties: plastic modulus, ultimate strain, and energy absorption (which is strongly dependent on plastic deformation). In contrast, the tensile properties of tibial tissue did not exhibit any age-related changes except in ultimate strain and energy absorption. Comparison by t test between the mechanical properties of tibiae from the third and ninth decades did not show significant differences ($p > 0.05$) except in ultimate strain and energy ab-

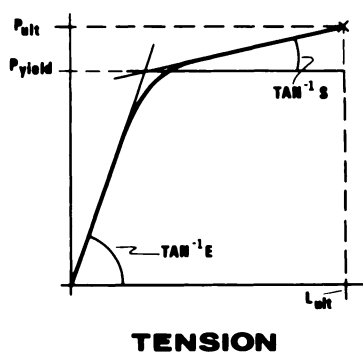


FIG. 1

Reduction method for tension load-deflection curves. P_{yield} was taken as the tension load at the start of yielding (initiation of plastic deformation). P_{uft} was the tension load at the point of fracture. The yield stress and ultimate stress in tension were found by dividing respective loads by the cross-sectional area of each specimen. L_{uft} was the elongation of the specimen at the point of fracture. The ultimate strain was found by dividing this elongation by the length of the specimen (gauge length). E , the elastic modulus, is found by multiplying the initial slope of the load-deformation curve by the ratio: gauge length/cross-sectional area. S , the plastic modulus, is found by multiplying the final slope by the same ratio.

TABLE
TENSION*

Age (Yrs.)	N	n	Femur					Energy (MN/m ²)
			Yield Stress (MN/m ²)†	Ultimate Stress (MN/m ²)	Elastic Modulus (10 ³ MN/m ²)	Plastic Modulus (10 ³ MN/m ²)	Ultimate Strain	
20-29	5	36	120 (6.4)	140 (10.0)	17.0 (2.24)	0.754 (0.1069)	0.034 (0.0067)	3.85 (1.104)
30-39	2	11	120 (9.9)	136 (3.5)	17.6 (0.28)	0.637 (0.0007)	0.032 (0.0092)	3.55 (0.976)
40-49	3	14	121 (8.4)	139 (10.7)	17.7 (4.45)	1.000 (0.2942)	0.030 (0.0040)	3.19 (0.525)
50-59	6	37	111 (11.9)	131 (12.6)	16.6 (1.74)	0.894 (0.1873)	0.028 (0.0059)	2.84 (0.614)
60-69	8	42	112 (5.9)	129 (6.4)	17.1 (2.21)	0.980 (0.1454)	0.025 (0.0055)	2.65 (0.781)
70-79	7	28	111 (6.5)	129 (5.5)	16.3 (1.78)	1.006 (0.2010)	0.025 (0.0060)	2.57 (0.678)
80-89	2	10	104 (5.0)	120 (7.1)	15.6 (0.71)	1.078 (0.3288)	0.024 (0.0021)	2.23 (0.120)
Per cent of change per decade*			-2.2	-2.1	-1.5	+8.0	-5.1	-6.8
Correlation coefficient			-0.910	-0.929	-0.770	+0.826	-0.979	-0.987

* The 20-29 decade was taken as the base period for all calculations. Data for each decade given as the means (S.D.) of the mean values for each of the bones tested, *N* being the number of bones tested and *n*, the number of machined specimens obtained from them.

† MN/m² = meganewtons per square meter.

sorption. The correlation coefficients confirmed that the only observed difference that could be attributed to age was the difference in ultimate strain. Although the energy change, based on mean decade values, was -8.4 per cent per decade and the correlation coefficient (*r*) was -0.80, the only statistically significant change in energy absorption appeared in the ninth decade. This may have been due to the small sample from the ninth decade (two individuals).

Six of the bones studied were from individuals who had been treated before death with corticosteroids for varying lengths of time. Four of the six were femora: the first from a sixty-nine-year-old man who had received steroids for three months for bronchogenic disease; the second from a sixty-eight-year-old man who had received steroids for one year for silicosis; the third from a thirty-two-year-old woman who had had steroid therapy for four years for asthma; and the fourth from a forty-one-year-old woman who had had steroid therapy for two years for lymphosarcoma. The two tibiae were from a fifty-seven-year-old woman with renal failure and rheumatoid arthritis who had received steroids for three years and a fifty-two-year-old man who had been treated with steroids for one year for bronchogenic carcinoma. In addition, there were two femora from individuals who were classified by their physicians as "osteoporotic," one from a fifty-five-year-old

woman who had had hormonal therapy for one year for a breast neoplasm with no metastases, the other from a seventy-five-year-old woman with bronchopneumonia.

We decided to compare these six femora and two tibiae with those in their age-matched population. However, since these "osteoporotic" bones had rather thin cortices, only enough specimens were available to perform one type of test. Therefore only tension tests were done. No significant differences were found in the means for the fourth, fifth, sixth, seventh, and eighth decades when these individuals were included or excluded.

No consistent differences could be found between the femora from the two "osteoporotic" individuals and those of the age-matched specimens. The femur from the fifty-five-year-old woman showed a significantly (*p* < 0.05) increased ultimate strain when compared with the mean of the other bones in the same decade but no significant difference in the other determinations. The femur from the seventy-five-year-old woman, however, sustained higher yield stress and ultimate stress (*p* < 0.01) and was stiffer, with a significantly higher elastic modulus (*p* < 0.001), than the mean of the other bones from the same decade.

The bone samples from the steroid-treated individuals showed no consistently different properties when compared with the means of the bones of normal individuals in the same decades. For example, the femoral specimens from the sixty-eight-year-old man with silicosis showed a significantly higher yield stress and ultimate stress (*p* < 0.001), a lower elastic modulus (*p* < 0.005), and a higher plastic modulus and ultimate strain (*p* < 0.005) than the mean values for the femora from the same decade, while the femoral specimens from the forty-one-year-old woman with lymphosarcoma showed a higher elastic modulus (*p* < 0.01) compared with the femora of other individuals in the same decade.

Compression Properties

Compression tests were performed on nineteen fem-

TABLE II
COMPARISON OF TENSILE PROPERTIES OF TIBIA AND FEMUR
AS SHOWN BY RATIOS OF TIBIAL TO FEMORAL VALUES*

	Yield Stress	Ultimate Stress	Elastic Modulus	Plastic Modulus	Ultimate Strain
\bar{x}	1.12	1.17	1.32	1.33	1.08
<i>S_x</i>	0.008	0.126	0.309	0.278	0.237
<i>N</i>	24	24	24	24	24
<i>P</i>	< 0.001	< 0.001	< 0.001	< 0.001	> 0.05

* \bar{x} = mean tibia-to-femur ratio for each property; *S_x* = standard deviation; *N* = number of bones compared.

I

N	n	Tibia					
		Yield Stress (MN/m ²)	Ultimate Stress (MN/m ²)	Elastic Modulus (10 ³ MN/m ²)	Plastic Modulus (10 ³ MN/m ²)	Ultimate Strain	Energy (MN/m ²)
3	19	126 (8.5)	161 (3.7)	18.9 (3.99)	1.17 (0.305)	0.040 (0.0107)	4.36 (0.582)
1	6	129 (—)	154 (—)	27.0 (—)	0.91 (—)	0.039 (—)	5.77 (—)
4	12	140 (4.2)	170 (6.5)	28.8 (9.19)	1.39 (0.218)	0.029 (0.0044)	4.09 (1.009)
6	23	133 (7.9)	164 (9.4)	23.1 (4.27)	1.21 (0.169)	0.031 (0.0016)	4.19 (0.384)
9	44	124 (8.7)	147 (9.2)	19.9 (2.44)	1.20 (0.127)	0.027 (0.0037)	3.05 (0.694)
4	16	120 (6.8)	145 (17.3)	19.9 (2.10)	1.18 (0.076)	0.027 (0.0084)	3.27 (1.606)
1	3	131 (—)	156 (—)	29.2 (—)	1.43 (—)	0.023 (—)	2.96 (—)
		-0.5	-1.2	+1.5	+3.4	-6.9	-8.4
		-0.22	-0.48	+0.13	+0.51	-0.93	-0.80

TABLE III
COMPRESSION

Age (Yrs.)	N	n	Femur		N	n	Tibia	
			Ultimate Stress (MN/m ²)	Elastic Modulus (10 ³ MN/m ²)			Ultimate Stress (MN/m ²)	Elastic Modulus (10 ³ MN/m ²)
20-29	2	11	209 (3.5)	18.1 (0.28)	—	—	—	—
30-39	2	9	209 (8.5)	18.6 (0.14)	1	3	213	35.3
40-49	2	12	200 (17.0)	18.7 (1.48)	2	6	204 (7.6)	30.6 (11.05)
50-59	4	18	192 (16.8)	18.2 (0.61)	2	8	192 (0.5)	24.5 (1.05)
60-69	5	27	179 (14.9)	15.9 (0.68)	4	14	183 (6.0)	25.1 (1.12)
70-79	3	14	190 (19.6)	18.0 (1.86)	1	4	183	26.7
80-89	1	4	180	15.4	1	3	197	25.9
% change per decade			-2.5	-2.2			-2.0	-4.7
Correlation coefficient			-0.906	-0.699			-0.68	-0.75

ora and eleven tibiae, five of the femora and five of the tibiae being paired bones from five individuals. The ultimate stress and elastic modulus by decade, as determined in these compression tests, are given in Table III. These two compression properties, both for the tibia and for the femur in each of the five pairs tested, were compared in the same manner that was used for the tensile tests (Table IV). As was the case with the tensile properties, the compression properties of the tibial tissue were significantly greater than those of the femoral tissue. The two bones were therefore considered separately. The femur showed significant ($p < 0.05$) decreases in the ultimate stress between the third and eighth and third and ninth decades. The correlation coefficient ($r = 0.906$) supports the assertion that this change was age-related. A comparison of these same decade groups showed no significant changes with age in the elastic modulus.

The tibial tissue, as in the tension tests, showed no significant changes with age in compression properties. Although the regression analysis indicated a trend in the eleven bones tested between the fourth and eighth decades ($r = -0.68, -0.75$), this was not significant.

Shear Properties

Torsion tests were performed on thirty-four tibiae and twenty-five femora, twenty femora and twenty tibiae being

paired bones from twenty individuals. The torsion tests on the twenty paired tibia-femur sets showed that the tibia-to-femur shear modulus ratios were 1.00 ± 0.05 . Therefore data for both bone sets were combined into Table V. Although a linear regression analysis of the mean decade values showed $r = -0.855$ for a -4 per cent change per decade, we could demonstrate no significant difference between the moduli of any of the age groups.

Discussion

Our findings that the mechanical properties of tibial bone tissue are greater than the corresponding properties of femoral bone tissue are consistent with previously presented data on ultimate strength of bone^{4,7,12}. The real value of our data is that they not only allow comparison of

TABLE IV
COMPARISON OF COMPRESSIVE PROPERTIES OF TIBIA AND FEMUR
AS SHOWN BY RATIOS OF TIBIAL TO FEMORAL VALUES*

	Ultimate Stress	Elastic Modulus
\bar{x}	1.12	1.67
S_x	0.064	0.278
N	5	5
P	< 0.02	< 0.01

* See Table II for explanation of terms.

TABLE V
SHEAR MODULUS: TIBIA AND FEMUR

Age (Yrs.)	N	n	Shear Modulus
20-29	4	27	3.58 (0.331)
30-39	3	27	3.58 (0.195)
40-49	6	37	3.70 (0.376)
50-59	15	93	3.23 (0.336)
60-69	14	86	3.10 (0.333)
70-79	14	88	3.17 (0.379)
80-89	3	15	3.17 (0.339)
% change per decade			-4.0
Correlation coefficient			-0.855

two bones under similar handling and testing conditions at a given age in one individual, but they also show differences in the two bones as functions of age. Also, these are the first data on the plastic properties of a population. With increasing age, femoral tissue undergoes a progressive, but slow, degradation in its mechanical properties with a concomitant increase in the slope of the plastic portion of its stress-strain curve. Tibial tissue, on the other hand, shows no regular change in mechanical properties with age except for ultimate strain and therefore the energy absorbed to fracture. Hence, femoral and tibial tissues differ in their mechanical properties at any one time and in their response to the aging process.

We believe that these differences in mechanical properties of the tibia and femur may represent differences between the two bones at the tissue and structure or organ level. At the tissue level, quantitative and qualitative differences in mineral and collagen could account for the different properties and differing changes with age. At the structural or organ level, the femur and tibia are subjected to differing load conditions and perhaps to different turnover rates *in vivo*. A higher turnover rate in the tibial

bone might account for its lack of age-related changes. Thus, if some age-related changes do occur after bone tissue has been formed, remodeling would remove and replace this aged bone and hence reduce such changes. An example of this phenomenon might be the consistency of the slope in the plastic zone for tibial bone (a reflection of its relatively new collagen) and the increase in plastic slope of the femoral bone, a reflection of its slower remodeling with persistent old collagen. The increased stiffness (slope) of femoral bone tissue in the plastic zone we believe to be a reflection of the aging of collagen. Recent investigation by our group shows that the slope in the plastic region of the stress-strain curve is dependent on the structure of the bone collagen³.

The most important age change that occurs in bone tissue, from a structural point of view, is the decrease in the plastic strain and hence the total or ultimate strain before failure. The decrease in ultimate strain in tension of 5 per cent in femoral bone and of 7 per cent in tibial bone per decade was mainly responsible for the 32 per cent and 42 per cent decreases in tensile energy absorption in the femoral and tibial tissue, respectively, between the third and ninth decades. That this decrease with age was seen in both femoral and tibial bone tissue suggests that the ultimate strain reflects a property or quality which depends on the age of the individual at the time that the bone tissue under test was originally deposited. Hence, age effects on ultimate strain are determined by the age of the organism rather than the time that the tissue has been *in situ*, as is the case with the plastic modulus.

Vose and co-workers, using embalmed femoral tissue specimens in a bending test, found that osteoporotic femora produced specimens of bone tissue which were stronger than those from normal femora. Our data on two osteoporotic individuals do not contradict this finding.

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